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SPINE

Biomechanics of coupled motion in the cervical spine during simulated whiplash in patients with pre-existing cervical or lumbar spinal fusion

A FINITE ELEMENT STUDY

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Objectives
Loss of motion following spine segment fusion results in increased strain in the adjacent motion segments. However, to date, studies on the biomechanics of the cervical spine have not assessed the role of coupled motions in the lumbar spine. Accordingly, we investigated the biomechanics of the cervical spine following cervical fusion and lumbar fusion during simulated whiplash using a whole-human finite element (FE) model to simulate coupled motions of the spine.

Methods
A previously validated FE model of the human body in the driver-occupant position was used to investigate cervical hyperextension injury. The cervical spine was subjected to simulated whiplash exposure in accordance with Euro NCAP (the European New Car Assessment Programme) testing using the whole human FE model. The coupled motions between the cervical spine and lumbar spine were assessed by evaluating the biomechanical effects of simulated cervical fusion and lumbar fusion.

Results
Peak anterior longitudinal ligament (ALL) strain ranged from 0.106 to 0.382 in a normal spine, and from 0.116 to 0.399 in a fused cervical spine. Strain increased from cranial to caudal levels. The mean strain increase in the motion segment immediately adjacent to the site of fusion from C2-C3 through C5-C6 was 26.1% and 50.8% following single- and two-level cervical fusion, respectively ($p = 0.03$, unpaired two-way *t*-test). Peak cervical strains following various lumbar-fusion procedures were 1.0% less than those seen in a healthy spine ($p = 0.61$, two-way ANOVA).

Conclusion
Cervical arthrodesis increases peak ALL strain in the adjacent motion segments. C3-4 experiences greater changes in strain than C6-7. Lumbar fusion did not have a significant effect on cervical spine strain.

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Keywords: Whiplash, Cervical spine, Coupled motions, Adjacent segment disease, Finite element

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Article focus
- A finite element model of the complete spine was used to assess biomechanics after surgery
- Coupled motions including multi-level cervical and thoracolumbar fusions were assessed using during realistic whiplash scenarios

Key messages
- Cervical fusion increases ALL strain at the adjacent segment in a realistic whiplash scenario.
- C6-C7 experienced the highest strain under these conditions, but cervical fusion had minimal effect. C3-C4 had a disproportionately high increase in strain following C4-C5 fusion.

- Thoracolumbar fusion had no effect on cervical spine strain.

Strengths and limitations

- This study determined that adjacent segment strain following spinal fusion disproportionately affects certain motion segments.
- A quantitative rationale for adjacent segment disease as a biomechanical consequence of fusion as well as adjacent segment disease as natural progression of underlying disease was established.
- A standardized 75-kg adult male and standardized Euro NCAP whiplash scenario was modelled. Other scenarios were not evaluated by this model.

Introduction

Whiplash injuries, caused by a sudden or unexpected neck flexion or extension, are the most often reported injuries in low-velocity rear-impact vehicle collisions.[1] The economic impact of whiplash is estimated to be \$3.9 billion annually in the United States, or more than \$29 billion when litigation costs are considered.[2] Some clinical challenges of whiplash include associated secondary gain that may confound injury,[2] associated non-organic signs of disability,[2] and the difficulty of establishing consistent diagnostic MRI findings.[3] Nonetheless, large defined population studies have shown that the incidence of patients presenting with whiplash related complaints may increase even when a concomitant decrease in insurance claims is observed.[4] Additionally, the overall body of literature provides evidence supporting a lesion-based model of whiplash injury.[5] Injury to the anterior longitudinal ligament (ALL) appears to be a marker to severe whiplash injuries in both cadaveric studies[6] and post-mortem studies.[7,8]

Most studies on whiplash have focused on patients with no pre-existing cervical spine disease. Cervical spine arthrodesis is a common procedure, with over 1.1 million patients undergoing the procedure over an eight-year period in the United States.[9] The effects of cervical arthrodesis on the adjacent motion segments above and below the level of fusion still remain a point of debate. It is clear that there is increased strain in the soft tissue and vertebrae in adjacent motion segments, due to compensation for the new loss of flexibility.[10,11] However, motion-sparing procedures, such as cervical disc arthroplasty, have failed to show decreased rates of adjacent segment disease (ASD), at least for single level surgery.[12]

We have previously quantified the biomechanical effects of an 8 g whiplash following cervical arthrodesis on adjacent segment strains seen in the ALL using a validated finite element (FE) model.[11] In our original study, the impact scenario selected was based on whole cadaver experiments performed by Mertz[13] in 1967 and whole cervical spine model experiments performed by Ivancic[14] in 2004, reflecting an 8 g peak acceleration. Although those historical impact pulses provided a method for validation and assessment of failure, newer data is available that reflects more realistic real-world, rear-impact collisions.[1] We sought to study cervical spine biomechanics after spinal fusion using the low-speed acceleration pulses described by the European New Car Assessment Programme (Euro NCAP). Additionally, our prior study used a simplified FE model that did not capture the additional effects caused by coupled motions from the lumbar spine, nor the effects created by the interaction between the model and the driver's seat. Advances in computational resources have allowed us to address these limitations.

We sought to provide quantitative data on the potential effects of adjacent segment strains following cervical and lumbar arthrodesis and test the hypothesis that specific motion segments experienced disproportionate changes in ALL strain and that there was no effect of lumbar fusion on cervical spine biomechanics.

Materials and Methods

Finite element model overview. THUMS (Total Human Model for Safety (Occupant Model v1.61, Toyota Central Research & Development Labs, Nagakute, Japan)), a 3D FE model of a 75 kg, 175 cm tall, 35-year-old male seated in a driving position, was used as the baseline for our study. THUMS incorporates roughly 91 200 total elements, 7000 of which represent the cervical spine, and 15 500 of which represent the remaining thoracic and lumbar spine. The spongy and cortical bone of the vertebral bodies were modelled as rigid solid and shell entities, respectively. The intervertebral discs were modelled using a combination of solid (pulposus and annulus) and seatbelt (lamellae) elements with isotropic material and section properties. The ALL, posterior longitudinal ligament (PLL), and other ligaments, including the ligamentum flavum, ligamentum nuchae, and intertransverse ligament, were represented using 2D shell elements, assigned with varying unique material properties. Only the deep layers of the ALL and PLL were modelled. All solid and shell elements in the cervical spine were defined as isotropic, linear elastic materials. Muscles were represented using Hill-type muscle models each comprised of two non-linear spring elements, a contractile element, and a viscous damper to address muscle viscoelasticity. Muscles and ligaments were allowed to react freely to applied loads, but no active muscle and ligament forces were incorporated into the model. Our model has been previously validated as a computational tool to quantify both local and global spine kinematics and validated for measurement of peak ALL strain under the tested loading conditions.[11,15]

All FE calculations were run using single-precision LS-DYNA R701 64-bit (Livermore Software Technology

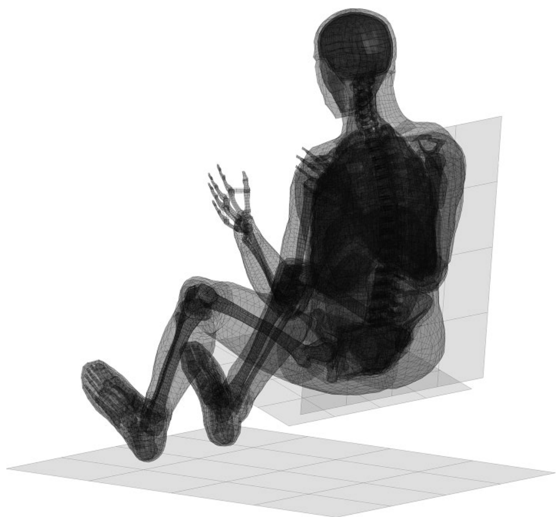

Fig. 1

Image of the validated FE model, with a rigid seat and floor shown using a semi-transparent filter.

Corp., Livermore, California) on a Microsoft Windows workstation (Core i5-3570k "Ivy Bridge", Intel Corp., Santa Clara, California; Z77 Extreme4 ATX, ASRock Inc. Taipei, Taiwan; BLS8G3D1609DS1S00, Micron Technology Inc., Boise, Idaho). LS-PrePost 3.2 64-bit (Livermore Software Technology Corp), Excel 2013 (Microsoft Corp., Redmond, Washington), and MATLAB 7.12.0.635 64-bit (MathWorks Inc., Natick, Massachusetts) were used to pre-process, post-process, and analyze the data, including determination of ALL strain. GraphPad 6.0 (GraphPad Software, La Jolla, California) was used for statistical analysis. Unpaired two-way Student's *t*-tests were used to compare change in adjacent segment average ALL strain following different cervical arthrodesis procedures. Paired two-way Student's *t*-tests were used to compare ALL strain and change in ALL strain in specific motion segments following different modelling conditions. A two-way analysis-of-variance (ANOVA) with Tukey post-test was used to compare change in cervical ALL strain following different thoracolumbar arthrodesis procedures.

Virtual surgery overview

All cervical and lumbar fusions were assumed to be complete and fully healed. This was simulated by rigidly constraining the cortical and spongy elements of adjacent vertebral bodies to one another. Instrumentation was not simulated.

Single-level arthrodesis was modelled by constraining two vertebral bodies as a single rigid entity. Two-level cervical arthrodesis assumed contiguous levels. ALL strains were evaluated at the neighbouring adjacent motion segments one and two levels away from the cervical fusion site. For instance, in the C4-C5 fusion case, the segments one level away were defined as ALL segments C3-C4 and C5-C6. The segments two levels away are defined as ALL segments C2-C3 and C6-C7. THUMS models the sacral spine as a fused structure; as such, L5-S1 arthrodesis was equivalent to arthrodesis of L5 to the sacrum.

All fusions were performed without alteration of the sagittal alignment of the model.

Loading conditions. Whiplash exposures were simulated by applying a 16 km/h ΔV, 10 g peak acceleration, 92 ms duration, triangular-shaped load curve to the model as defined by Euro NCAP and the International Insurance Whiplash Prevention Group (IIWPG) to represent whiplash-inducing accidents.[1,16] Loads were applied to the model in two different ways: 'T1 Acceleration' and 'Seat Interaction'.

T1 Acceleration Model. In the first set of simulations, all nodes at or below the T1 vertebrae in the FE model were treated as a rigid body and loaded in the posterior to anterior x-direction. The load was removed after 92 ms and the model was allowed to continue its inertial movement until the termination of the simulation at 300 ms. This simulates the cervical spine and head in isolation, and omits the contribution of the coupled motions of the lumbar spine. This most closely reproduces the cadaveric whole cervical spine model with muscle force replication developed by Ivancic and Panjabi.[14]

Seat Interaction Model. In the second set of simulations, a seat and floor were added using rigid 4-by-4 element shells through LS-PrePost. The seat angle was fit to the curvature of our validated FE model, and in accordance with the angles described in SAE J826 H-point mannequins fitted with a Head Restraint Measuring Device and the Euro NCAP whiplash testing protocol.[17,18]

Initially, the seat and floor were fixed in all six degrees of freedom (x-, y-, z-translational and x-, y-, z-rotational) and positioned 1 mm below the THUMS model. Gravity was then applied for three seconds to relax the THUMS model into the seat in preparation for the ensuing whiplash exposure. Whiplash was again simulated as described above. The FE model with the added seat and floor is shown in Figure 1.

Results

Representative snapshot images of the whiplash simulation are in Figure 2. Figure 3 compares the differences in resultant head acceleration between the T1 Acceleration Model and the Seat Interaction model. The T1 Acceleration Model showed a 10 g increase and 8 ms phase delay in peak head acceleration.

T1 Acceleration Model. An increase in ALL strain was seen for all fusion conditions (Table I). ALL strain in the baseline model ranged from 0.081 to 0.304, increasing from cranial to caudal levels. With the exception of C6-C7, ALL strain increased at all levels and ranged from 0.085 to 0.309. The C6-C7 motion segment experienced the smallest mean increase in strain (0.58%,

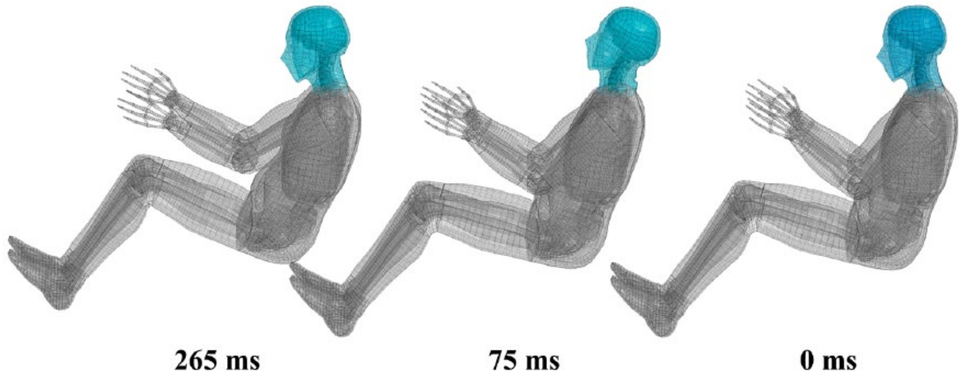

Fig. 2a

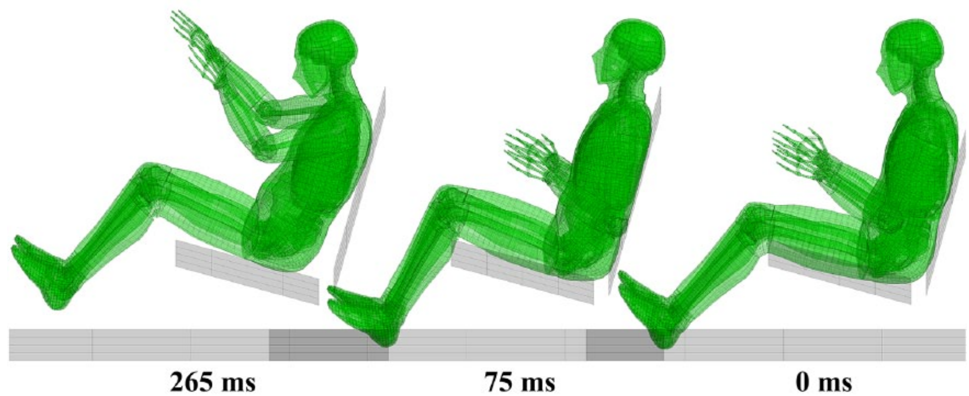

Fig. 2b

Snapshots of the a) T1 Acceleration Model and b) Seat Interaction Model. At 0 ms in the T1 Acceleration Model, a 16 km/h pulse with a 10 g peak acceleration is applied leftward to all dummy nodes at or below the T1 vertebrae (grey). By 75 ms, most of the load has been applied. By 265 ms, all of the load has been applied and the dummy is moving forward at a constant velocity. At 0 ms in the Seat Interaction Model, the same pulse is applied leftward to the rigid seat (grey). At 75 ms, the seat is slightly past mid-impact with the dummy. By 265 ms, the impact is complete resulting in the dummy separating from the seat.

Table I. T1 Acceleration Model: Peak cervical spine anterior longitudinal ligament (ALL) strain and percentage increase following cervical fusion. Peak strain of the ALL increased following spinal fusion. The difference in adjacent segment mean strain in C2-C3 to C5-C6 between single- and two-level fusion was significant (p = 0.019, two-way unpaired *t*-test)

		Single-level fusion, ALL strain (% increase)					**Two-level fusion, ALL strain (% increase)**			
ALL segment	**Baseline**	**C2-C3**	**C3-C4**	**C4-C5**	**C5-C6**	**C6-C7**	**C2-C4**	**C3-C5**	**C4-C6**	**C5-C7**
C2-C3	0.081	N/A	0.092 *(13.6)*	0.099 *(22.2)*	0.091 *(12.3)*	0.085 *(4.9)*	N/A	0.126 *(55.6)*	0.122 *(50.6)*	0.099 *(22.2)*
C3-C4	0.117	0.133 *(13.7)*	N/A	0.172 *(47.0)**	0.148 *(26.5)*	0.150 *(28.2)*	N/A	N/A	0.241 *(106.0)**	0.178 *(52.1)*
C4-C5	0.169	0.187 *(10.7)*	0.201 *(18.9)*	N/A	0.22 *(30.2)*	0.196 *(16.0)*	0.223 *(32.0)*	N/A	N/A	0.246 *(45.6)*
C5-C6	0.180	0.187 *(3.9)*	0.197 *(9.4)*	0.232 *(28.9)*	N/A	0.234 *(30.0)*	0.208 *(15.6)*	0.277 *(53.9)*	N/A	N/A
C6-C7†	0.304	0.304 *(0.0)*	0.304 *(0.0)*	0.306 *(0.7)*	0.309 *(1.6)*	N/A	0.305 *(0.3)*	0.308 *(1.3)*	0.322 *(5.9)*	N/A

*The ALL at C3-C4 following C4-C5 fusion and C4-C6 fusion had the greatest change in strain for single- and two-level fusion, respectively
†Although absolute strain was highest as C6-C7, this level experienced the least amount of change following fusion

SD 0.78%) and seemed relatively unaffected, even by adjacent level fusion. The C3-C4 segment saw the greatest change in mean strain of 47% following C4-C5 fusion. In the motion segments from C3-C6, the greatest increase in strain was seen with a single level fusion at the level immediately below the site of fusion. That is, C3-C4 ALL strain was greater following C4-C5 fusion as opposed to C2-C3 fusion, and C4-C5 ALL strain was greater following C5-C6 fusion as opposed to C3-C4 fusion. Following two-level cervical fusion, ALL strain ranged from 0.099 to 0.322. At individual motion segments, the C6-C7 motion segment saw the lowest mean change in strain (2.5%, SD 3.0%) while again, C3-C4 saw the greatest mean increases in strain (79.1%, SD 38.1%). When analyzed in aggregate, the mean change in adjacent segment strain was 23.0% (SD 13.9%) for

Table II. Seat Interaction Model: Peak cervical spine anterior longitudinal ligament (ALL) strain and percentage increase following cervical fusion. Peak strain of the ALL increased following spinal fusion. Absolute strain was generally higher in the Seat Interaction Model compared to the T1 Acceleration Model. The difference in adjacent segment strain in C2-C3 to C5-C6 between single- and two-level fusion was significant (p = 0.03, two-way unpaired *t*-test)

		Single-level fusion, ALL strain (% increase)					Two-level fusion, ALL strain (% increase)			
ALL segment	**Baseline**	**C2-C3**	**C3-C4**	**C4-C5**	**C5-C6**	**C6-C7**	**C2-C4**	**C3-C5**	**C4-C6**	**C5-C7**
C2-C3	0.106	N/A	0.132 *(24.5)*	0.127 *(19.8)*	0.122 *(15.1)*	0.116 *(9.4)*	N/A	0.170 *(60.4)*	0.142 *(34.0)*	0.132 *(24.5)*
C3-C4	0.145	0.162 *(11.7)*	N/A	0.207 *(42.8)**	0.194 *(33.8)*	0.157 *(8.3)*	N/A	N/A	0.264 *(82.1)**	0.233 *(60.7)*
C4-C5	0.202	0.224 *(10.9)*	0.223 *(10.4)*	N/A	0.264 *(30.7)*	0.197 *(-2.5)*	0.255 *(26.2)*	N/A	N/A	0.270 *(33.7)*
C5-C6	0.209	0.225 *(7.7)*	0.243 *(16.3)*	0.289 *(38.3)*	N/A	0.260 *(24.4)*	0.272 *(30.1)*	0.317 *(51.7)*	N/A	N/A
C6-C7†	0.382	0.384 *(0.5)*	0.382 *(0.0)*	0.387 *(1.3)*	0.398 *(4.2)*	N/A	0.387 *(1.3)*	0.398 *(4.2)*	0.399 *(4.5)*	N/A

*The ALL at C3-C4 following C4-C5 fusion and C4-C6 fusion had the greatest change in strain for single- and two-level fusion, respectively
†Although absolute strain was highest as C6-C7, this level experienced the least amount of change following fusion

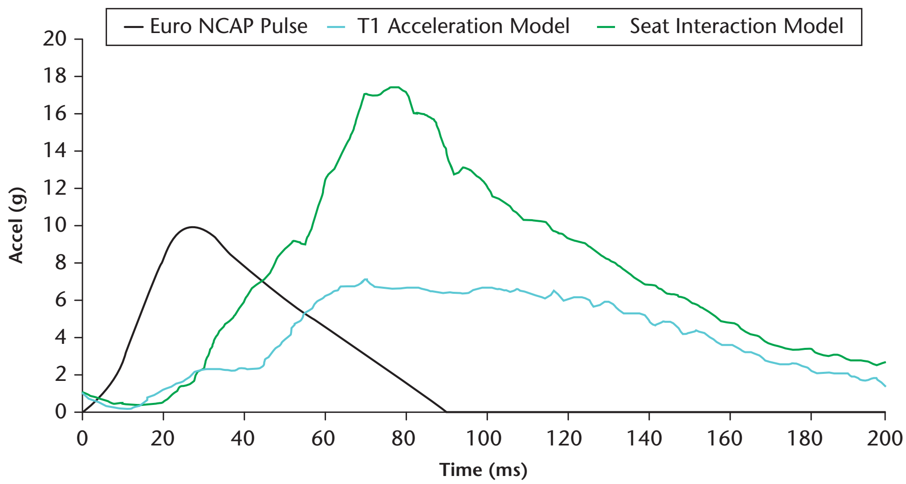

Fig. 3

Comparison of resultant head accelerations in two models loaded using the same Euro NCAP pulse. The T1 Acceleration Model experienced a 7.2 g peak head acceleration at 70 ms. The Seat Interaction Model experienced a 17.5 g peak head acceleration at 78 ms. A 5-point averaging filter was applied to the Seat Interaction Model curve to smooth out short-term fluctuations.

single-level fusion and 49.8% (SD 33.1%) for two-level fusion (p = 0.056, two-way unpaired *t*-test). If C6-C7 is removed from the statistical analysis, the mean change in adjacent segment strain in C2 through C6 was 26.0 (SD 11.8%) for single-level fusion and 58.6% (SD 28.1%) for two-level fusion (p = 0.019, two-way unpaired *t*-test).

Peak strain of the ALL increased following spinal fusion. The ALL at C3-C4 following C4-C5 fusion and C4-C6 fusion had the greatest change in strain for single- and two-level fusion, respectively. Although absolute strain was highest as C6-C7, this level experienced the least amount of change following fusion. The difference in adjacent segment mean strain in C2-C3 to C5-C6 between single- and two-level fusion was significant (p = 0.019, two-way unpaired *t*-test).

Seat Interaction Model. An increase in ALL strain was seen under all conditions. (Table II). ALL strain ranged from 0.106 to 0.382 in our model, increasing from cranial to caudal levels. Following single-level fusion, ALL strain ranged from 0.116 to 0.398. At individual motion segments, the C6-C7 motion segment experienced the lowest mean change in strain (1.51%, SD 1.87%) following all tested single-level fusion conditions, while C3-C4 saw the greatest change of mean ALL strain 42.8% following C4-C5 fusion. In the motion segments from C2-C5, the greatest increase in strain was seen with a single-level fusion at the level immediately below the site of fusion. Following two-level cervical fusion, ALL strain ranged from 0.132 to 0.399. At individual motion segments, C6-C7 motion segment saw the lowest mean change in strain (3.3%, SD 1.7%) while again, C3-C4 saw the greatest mean increases in strain of 71.3% (SD 15.1%). When analyzed in aggregate, the mean change in adjacent segment strain following single-level fusion was 23.4% (SD 13.8%), and it was 43.1% (SD 27.4%) following two-level fusion (p = 0.10). If C6-C7 is removed from the statistical analysis, the mean change in adjacent segment mean strain in C2-C3 through C5-C6 following single-level fusion was 26.1% (SD 12.3%), and it was 50.8% (SD 22.2%) following two-level fusion (p = 0.03, unpaired two-way *t*-test).

Lumbar fusion had no meaningful impact in cervical spine strain, with a mean change of -1.0% strain (SD 4.1%) (Table III). No apparent pattern could be recognised when evaluating a single-level lumbar fusion to a complete thoracolumbar fusion (p = 0.61, ANOVA).

Table III. Seat Interaction Model: Peak cervical spine anterior longitudinal ligament (ALL) strain and percentage increase following lumbar fusion. Lumbar fusion had minimal impact on cervical spine strain and change in strain. The mean change for all tested conditions was -1.0% strain (SD 4.1%. There was no difference in change when comparing a single level lumbar fusion at L5-S1 with a thoracolumbar fusion from T9-S1 ($p = 0.61$, ANOVA)

ALL segment	Baseline	L5-S1 fusion	L2-L5 fusion	L2-S1 fusion	T9-S1 fusion
C2-C3	0.106	0.107 *(0.9)*	0.107 *(0.9)*	0.106 *(0.0)*	0.111 *(4.7)*
C3-C4	0.145	0.137 *(-5.5)*	0.134 *(-7.6)*	0.135 *(-6.9)*	0.136 *(-6.2)*
C4-C5	0.202	0.203 *(0.5)*	0.202 *(0.0)*	0.203 *(0.5)*	0.179 *(-11.4)*
C5-C6	0.209	0.215 *(2.9)*	0.212 *(1.4)*	0.214 *(2.4)*	0.209 *(0.0)*
C6-C7	0.382	0.382 *(0.0)*	0.384 *(0.5)*	0.385 *(0.8)*	0.385 *(0.8)*

Comparisons between models. Modelling the whole human with seat interaction generated a 23% mean increase in ALL strain (SD 9.5% for single-level fusion; SD 8.3% for two-level fusion; both $p < 0.0001$, unpaired two-way test). However, when comparing the mean change in strain following single-level fusion, adding the seat interaction did not result in a significant difference (-0.5%, SD 7.5%, $p = 0.76$; -0.2%, SD 10.1%, $p = 0.48$, paired two-way *t*-test).

Discussion

A validated complete human FE model was updated to investigate the effects of cervical and lumbar fusions on peak ALL strains seen in the neck, under a realistic whiplash impact pulse with the addition of a rigid car seat and floor. Based upon the FE analysis, cervical fusion increases the risk of injury at specific levels that may not have preexisting risk (i.e. C3-C4), but does not appear to dramatically increases peak ALL strains in portions of the spine that are already high risk (i.e. C6-C7). Additionally, while changes to motion at segments adjacent to a fusion can be measured, these changes are limited to the immediate segments (at most). Fusion at the thoracolumbar spine did not appear to affect the cervical spine in our model.

The inclusion of seat interaction and thoracolumbar coupled motions resulted in a 23% increase in the ALL strains ($p < 0.0001$), indicating that modelling the seat is important in injury prediction. There was no significant difference in the way the models assessed the changes in adjacent segment ALL strain following fusion, suggesting that the more computationally efficient T1 Acceleration Model may be adequate to quantify these effects.

Increase in ALL strain was seen following all cervical spine procedures. Statistically significant increases in ALL strain between one- and two-level fusions were seen in both the T1 Acceleration Model and Seat Interaction Model ($p = 0.019$ and $p = 0.03$, respectively) when analyzing the motion segments between C2-C6. C6-C7 showed high levels of ALL strain under all tested conditions including the baseline, non-operative condition. Our strain of 0.3 in the baseline model was close to the tolerance of the ALL according to Yoganandan,[19] but well below the 0.8 from the dynamic tests of Bass.[20] The addition of a seat back increased ALL strain by 23.0% ($p < 0.0001$). However, the percent change in ALL strain following surgery remained consistent for all conditions ($p = 0.76$ for single-level; $p = 0.48$ for two-level fusion). While absolute strain values varied, the consistent percent change suggests that our model is robust in its assessment in change of adjacent segment ALL strain following cervical fusion under a broad range of conditions (i.e. with and without a seat back, and EURO-NCAP acceleration vs Yoganandan's acceleration pulse). The increased strain in the whole-human test condition was likely a result of the additional shorter impulse delivered to the neck due to the decoupling of the body from the seat and the vertical acceleration component from straightening of the compliant lumbar and thoracic spines. Additionally, the presence of soft-tissue interactions surrounding the muscle elements may have further contributed to the larger strains to the cervical spine. Lumbar fusion had no meaningful effect on cervical strain (-1.0% change, $p = 0.61$).

Complete ALL failure has been documented to occur at threshold strains between 0.426 and 0.476,[19] with partial injury occurring at strains above 0.222.[14] Our results suggest partial injury during whiplash to the ALL at C6-C7 for a patient with a normal, unfused spine. Patients who have undergone fusion appear to have an increased risk of injury. In most of our simulations, one or more ALL segments would have reached partial failure thresholds following fusion.[14]

Lumbar fusion does not appear to have any impact in the cervical spine during whiplash. The claim that neck symptoms have been exacerbated by prior lumbar fusion in the absence of sagittal imbalance is not supported by this study.

Several important clinical conclusions can be made from this data. Modern reports of ASD indicate C4-C5 and C6-C7 as the levels most affected by ASD.[21] Plate distance from the vertebral endplate has also been shown to affect adjacent segment ossification.[22] Our data shows that C6-C7 experiences the highest strain at baseline, and that single-level cervical fusion did not appear to have meaningful difference in biomechanical strain. In contrast, C3-C4, which under normal conditions experiences lower ALL strain and is infrequently a treated motion segment in isolation, experienced the largest increases in cervical ALL strain following arthrodesis in comparison with the other motion segments.

Thus, our data is in agreement with contemporary thinking that ASD reflects both a natural history of spondylosis (C6-C7) as well as biomechanical factors (C3-C4). Perhaps motion-sparing surgery at C4-C5 will have advantages over fusion when considering adjacent disease at C3-C4.

Several limitations exist for future consideration but do not affect the conclusions drawn above. Our material properties were isotropic and did not account for strain rate variability. Although our FE model is validated for the acceleration rate simulated, the correlation between cadaveric and FE simulations may be different at different accelerations. Additionally, in our FE model, the ligaments were rigidly attached to vertebral bodies; there may be high strains at these attachments clinically. In real-world vehicle collisions, muscles contract either in anticipation or as an instantaneous response to the impact, which may exacerbate or alleviate these strains depending on the subject-dependent muscle activations.[23,24] Although a realistic whiplash impulse was used, our data cannot be applied directly in a forensic situation, as our seat is rigid and based on the geometry in the SAE J826 standard. Our seat did not include a headrest or incorporate whiplash prevention and mitigation technologies currently found in most vehicles. Additionally, the aetiology of whiplash syndrome may involve the facet capsules.[25,26] Unfortunately, the capsules are not modelled with sufficient fidelity in our model for a confident strain analysis. However, these limitations do not change the points made in this study. Finally, our computational model only represents at 50 percentile American male. Changes in height or weight may result in absolute differences in ALL strain. However, the conclusions drawn – specifically that peak ALL strain increases in the cervical spine adjacent to a fusion, and that the increase in strain disproportionately affects different motion segments – is unlikely to change when differently sized models are used. Importantly, the use of a standardised model ('AM50') allows better comparison with available literature and existing mechanical and cadaveric models.

In conclusion, this study quantifies the effects of whiplash neck injury in scenarios wherein the occupant has previously undergone cervical or lumbar arthrodesis. Peak ALL strains were found to be higher in the motion segments adjacent to the level of fusion compared with a healthy spine, and strains directly increased with longer fusions. C3-C4 was the motion segment at greatest biomechanical risk for adjacent segment strain following arthrodesis, while C6-C7 showed high strain levels regardless of arthrodesis status. Lumbar fusion had no statistical or clinically meaningful effect on ALL strains in the cervical spine. This biomechanical study suggests that the debate about whether adjacent segment disease is the effect of natural history or the effect of couple motions may require investigation of specific motion segments.

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Author Contribution
- H. Huang: Conceived and designed the study, Discussed the results and implications, Performed the computations and analyzed the output data. Contributed to the manuscript at all stages.
- R. W. Nightingale: Conceived and designed the study, Discussed the results and implications, Contributed to the manuscript at all stages.
- A. B. C. Dang: Conceived and designed the study, Discussed the results and implications, Contributed to the manuscript at all stages.

Conflict of Interest Statement
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